Array Coil Types and Design Principles

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Introduction

Before we embark on the topic of array coils designs, the reader is referred to a historic paper by Hoult [1] that provides a detailed overview and discussion on the NMR receiver. Most of today's RF receive coil designs are based upon the phased array coil technology proposed by Roemer et al. in 1990 [2], and its application to volume imaging was discussed by Hayes et al. in 1991 [3]. (For the details of original and general phased-array radar technology, the reader may be referred to [4, 5].) The reader is also encouraged to read a detailed review of array coil theory and application by Wright and Wald [6]. It is well known that a small surface coil yields higher SNR at the distances closer to the surfaces of the coil when compared with that of a volume coil such as birdcage coil [7]. On the other hand, a small surface coil B_1-sensitive region is (much) smaller than that of a volume coil. What is remarkable in the phased array coil approach is that the high SNR associated with a small surface coil can be achieved and maintained over an extended FOV, by using a set of RF coils, which may be comparable to that of a volume coil. This is enabled by the development of various array coil designs for imaging applications of interest and the techniques to decouple the mutual inductances and thus the crosstalk among the array coil elements. Adjacent coil elements are often partially overlapped to cancel the mutual inductance between the elements. Next nearest and other coil elements can be mutually decoupled by the use of low input impedance (typically less than 5 Ohm) preamplifier decoupling circuit (other decoupling techniques and methods are described briefly later in this paper). For detailed discussions of low noise amplifiers, the reader may refer to [8]. What the low input impedance preamplifier effectively does in conjunction with a coil matching/decoupling circuit is to eliminate the current flow in the coil element loop, thereby eliminating the magnetic fields induced in the neighboring coil elements. In designing array coils, baluns and management of cable routing are also important factors to consider, and they are discussed elsewhere in this educational course.

Recent advancements on MR scanners with more-available-receiver channels, together with the rapid developments of parallel imaging techniques [9, 10, 11], are pushing the envelope of RF coil technologies. An example is the 96-channel head array coil [12]. In particular, 1.5T and

3T MRI applications are now realized and accepted as everyday routine clinical practices. At both field strengths, the key requirements are maximum achievable SNR and the coil optimization to the clinical applications within the framework of the parallel imaging scheme. (Remember that SNR is destined to be degraded or lowered from the start point, the signal source, to the end point of the receive chain. Thus, the question is how to minimize the loss of SNR in that process.) In this paper, the components constituting an array coil are explained, along with remarks about issues that should receive special attention.

A Single Coil Model

A schematic representation of a single coil including a preamplifier decoupling circuit is shown in Figure 1. This coil circuit is a building block of an array coil. L_1 and R_1 represent the coil inductance and resistance (typically around 0.5 Ohm in air and approximately 5 Ohm when placed on a phantom), respectively. C_1 and C_2 are tuning and matching capacitances, respectively. L_2 is a matching inductor that plays an extremely important role; that is, a part of L_2 functions to match the coil impedance to 50 Ohm together with C_2 when looked at from the side of the preamplifier. At the same time, the remaining part of L_2 achieves a parallel resonant circuit formed with C_2, in particular, when the input impedance of the preamplifier, r_{preamp}, becomes small (say, 0 to 2 Ohms). What this means is that when looked at from the side of the coil, the impedance is infinity, thereby, a high-impedance or open circuit. In other words, **the decoupling circuit considered here is equivalently a** $\frac{\lambda}{4}$ **(quarter wavelength cable) circuit; one end open and the other end short, the important physical property of a** $\frac{\lambda}{4}$ **transmission line**. (In fact, a $\frac{\lambda}{4}$ cable is often used as an impedance transformer. For example, a T/R switching circuit also finds the use of a $\frac{\lambda}{4}$ cable using the properties of open circuit and cross diodes separating the transmit chain from the receive chain.) What is extremely important to note here is that the current cannot flow in an open coil which eliminates the possibility of producing any induced magnetic field through non-zero mutual inductances present among all the neighboring coil elements. This is the underlying principle that explains why the crosstalk among all coil elements can be eliminated. Incidentally, the Q of the coil (characterized by C_1, L_1 and R_1), when the decoupling circuit is in effect, remains unchanged. The important thing to note here is that the signal source is changed from the current source (without preamplifier decoupling) to the voltage source (with preamplifier decoupling) that contains all the necessary information without losing the integrity of the original signal information.

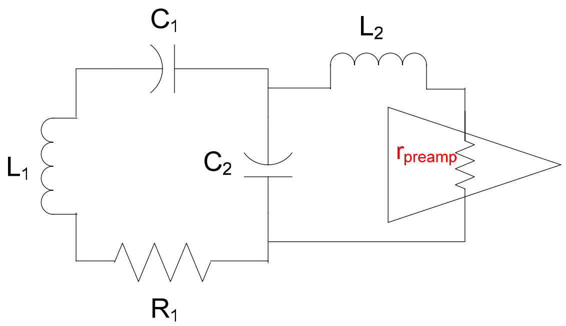

Figure 1 A single coil schematic representation with a low input impedance preamplifier

Signal-to-Noise Ratio

SNR is one of the most important parameters to be optimized in MRI applications, and its detailed discussion is given in [13]. Here, it is essential to understand how SNR relates to the coil-related parameters. The signal is written as

$$S \propto \gamma^3 B_0^2 B_1^{xy}(r) \qquad \text{Eq. (1)}$$

where γ is the nuclear gyromagnetic ratio, B_0 the static magnetic field, and $B_1^{xy}(r)$ the RF magnetic field produced by a coil with a unit current 1A. Thus, in designing a coil, $B_1^{xy}(r)$ must be optimized. What this means is that the coil size over a target FOV and the distance from the FOV need to be appropriately chosen to maximize $B_1^{xy}(r)$. The size to be chosen also depends upon the number of available receiver channels, in general.

On the other hand, NMR/MRI noise is thermal noise, and the noise generated from the coil is given by

$$N = \sqrt{4kT\Delta fR} \qquad \text{Eq. (2)}$$

where k is the Boltzmann constant, T the temperature in K, Δf the bandwidth, and $R=R_C+R_S$. It is noted that R_C is the coil resistance and R_S the sample energy loss (i.e., the equivalent series resistance due to the induced eddy current losses in the conductive sample). Combining Eqs. (1) and (2), SNR can be expressed by just using coil-related parameters as

$$\frac{S}{N} \propto \frac{B_1^{xy}(r)}{\sqrt{R_C + R_S}} \qquad \text{Eq. (3)}$$

To optimize SNR, $B_1^{xy}(r)$ can be maximized by having the coil closer to the sample, and R_S can be minimized by choosing the coil size to match the target FOV. R_C is minimized by making the unloaded coil Q high. It is well known that at low-B_0 field systems ($\leq$ 0.3T) the coil resistance is dominant or at least comparable to the sample noise, but at 1.5T and above, the sample noise dominates (i.e., $R_S >> R_C$). Furthermore, it is useful to be aware that if two coils yield the same relative sensitivity in free space and if they are each sample noise dominated, the two coils have the same absolute sensitivity [14]. This reference discusses the coil unloaded Q, loaded Q, and the sensitivity in detail, which should be useful to the coil engineer.

Low Noise Preamplifier

As seen previously, a low input impedance preamplifier plays a critical role in designing an array coil, and furthermore the characteristic parameters of preamplifiers such as noise figure certainly affect SNR. What a preamplifier does is as follows. The electromotive force or induced voltage (signal) in a coil is very small and typically on the order of a few μV. This small signal is amplified to a few mV by a preamplifier whose gain is, say, gain of 30dB (i.e., 1000 times greater; the reader is referred to Table 1 for quick conversion factors; it is also noted that the gain usually refers to power gain in practice, thus 20dB power gain means 10dB voltage gain).

Factor	1/100	1/10	1/5	1/3	1/2	2	3	4	5	6	7	10	100
dB	-20	-10	-7	-4.8	-3	3	4.8	6	7	7.8	8.45	10	20

Table 1 Relationship between dB expression and common factor

One of the parameters to measure the performance of preamplifier is the noise figure (NF) and discussed in detail elsewhere [1, 15]. Depicted in Figure 2 is a simple model for preamplifier noise.

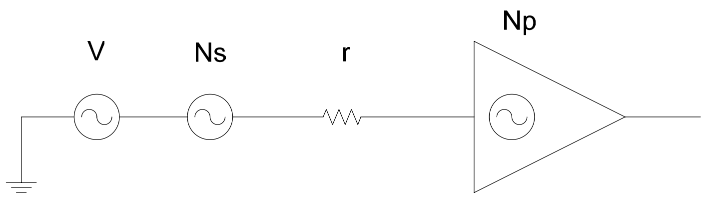

Figure 2 Depiction of a preamplifier noise model

In Figure 2, V, N_S, r and N_P denote the signal, the noise generated by the resistance r of the input signal source and the noise generated inside the preamplifier, respectively. Using these quantities, NF is defined as

$$NF = 10\log_{10}\left(\frac{N_P^2 + N_S^2}{N_S^2}\right) \quad \text{where} \quad N_S = \sqrt{4kT\Delta fr} \qquad \text{Eq. (4)}$$

A preamplifier is noise matched. What this means is that for a given FET there exists a signal source impedance that optimizes or minimizes the noise figure. Thus, the optimum noise figure can be achieved when the impedance of the receive coil matches with the source impedance of the FET. The FET impedance is relatively high (>>50 Ohm), and thus there exists some sort of impedance transformer (a part of L_2 in Figure 1 and a series capacitor grounded after L_2 , i.e., an LC circuit that you do not see in Figure 1) that transforms 50 Ohms to the input impedance of the FET. The industry standard preamplifier NF is currently less than 0.5 dB. It is also well known that the first NF and gain have the most significant impact in the entire electronics circuit which is often cascaded, as shown in low noise amplifier design textbooks such as [8, 15]. Noise is generated in any passive elements, including cables, in any electronics that dissipates power. This is the reason that most array coils have preamplifier integrated design to minimize undesirable noise contributions.

Array Coil Designs

Although the array coil designs employing the preamplifier decoupling are focused in this paper, it is useful for the reader to know different decoupling techniques. To name a few, they are:

- Overlap [2],
- Quadrature (intrinsic isolation/decoupling),
- Solenoidal array employing the anti-turn loop [16],
- Capacitor decoupling to cancel the inductance [17],
- Decoupling network after coils to manipulate the electric filed induced by the mutual inductive coupling [18], and
- Shield or transmission line designs because the most electromagnetic fields are contained between the coil element and the shield [19] (In this case, only some amount of electromagnetic fields leak out for imaging purpose. Even if the leaked electromagnetic fields couple with each other, the isolation between the coils is still adequate as the leaked fields are a small portion of the entire fields.)

Going back to the array coil designs employing the preamplifier decoupling, we have reviewed in the previous sections how each coil element is constructed to produce no crosstalk among all other coil elements. Figure 3 shows a 4-channel array coil to illustrate the concept.

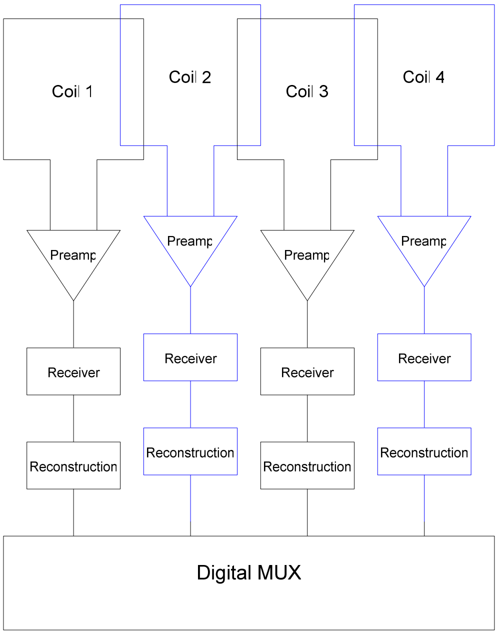

Figure 3 Schematic representation of an array coil (4-channel)

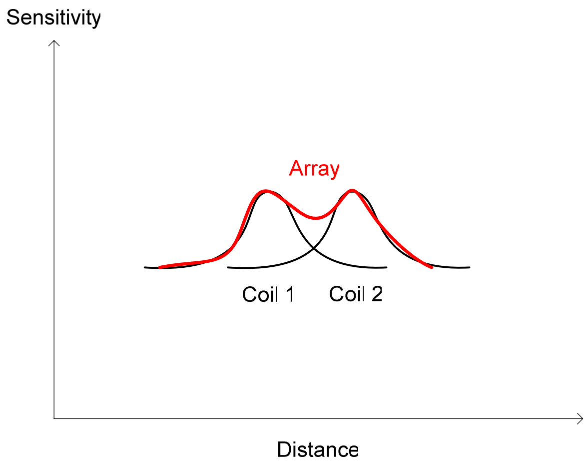

Figure 4 Depiction of B_1 sensitivity profile in an array coil

Each coil has its own B_1 sensitivity profile as shown in Figure 4. Because the coil is a surface coil, which yields a high SNR only over a small region, high SNR can be generated over a larger region of interest, i.e., FOV or ROI, if each channel SNR is combined by a Sum-of-Squares method [2], for example. This achieves the overall high SNR over the target FOV, shown in Figure 4.

It is straightforward to extend the array concept to 8-channel, 16-channel, and even larger numbers of channels. The number of the coil channels is determined by the number of available receiver channels and the target FOV or ROI while considering the appropriate size of each coil with SNR optimization in mind. There are many references that discuss different array coil designs for various imaging applications at different field strengths. Although an array coil using quadrature pairs (for each channel) is possible, the current trend is to use a set of various (size and shape) loop coils as the number of available receiver channels increases.

With the advent of parallel imaging techniques [9, 10, 11], the array coil designs play a major role in achieving the desired parallel-imaging performance. There are many references available for this discussion [20-31]. However, it is noted that there is a major difference between SMASH and SENSE with respect to the coil designs. SMASH is a ‘‘k-space’’ reconstruction approach whereas SENSE is an ‘’x-space’’ (i.e., image domain) reconstruction approach. In SMASH, spatial harmonics need to be formed by manipulating each coil B_1 sensitivity in the array to replace some k-space data lines which would have been otherwise collected by the application of conventional gradient phase encoding. This means that the direction of the array must coincide with the desired phase encoding direction. The human body has complex shapes and curvatures, and it is not always easy to construct a SMASH array coil whose direction/orientation coincides with a desired phase encoding direction in a routine clinical imaging practice. On the other hand, SENSE is less restrictive as to how each-channel coil is placed in conjunction with a chosen phase encoding direction. However, in SENSE, it is essential for one to be able to tell which B_1 sensitivity belongs to which coil in order to unfold the aliased image that is created and thus achieve a desired acceleration factor. The acceleration factor is the inverse of the reduction factor in imaging time. The measure of how ‘’easy’’ it is to tell the B_1 sensitivity distinctiveness is given by a so-called g-factor. Thus, the coil design work requires the optimization of the g-factor for a given imaging application [20]. Furthermore, the acceleration factor depends upon the number of receiver channels. That is, if you wish to achieve, for example, an acceleration factor of 3 in x-direction, there need to be at least 3 individual coils in that direction.

Summary

The research areas of array coil designs are indeed profound and broad from the perspective of the entire receive chain and the emerging applications of parallel imaging. There exist many coil designs for different applications at various B_0 field strengths. However, the underlying design principles for the array coil designs are essentially the same. What we must keep in mind when designing RF array coils may be summarized as follows:

(1) clinical applications of interest including the directions of phase encoding and the target

acceleration factor;

(2) the number of available receiver channels, the field strength and the desirable FOV; and

(3) optimization of the clinical applications of interest (SNR, uniformity, FATSAT, etc).

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